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Effect of Frontal Plane Foot Position on Lower Extremity Muscle Activation and Limb Positioning in a Single Leg Squat

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EFFECT OF FRONTAL PLANE FOOT POSITION ON LOWER EXTREMITY MUSCLE ACTIVATION AND LIMB POSITIONING IN A SINGLE LEG SQUAT

By

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This scholarly project, submitted by Brian Bolin in partial fulfillment of the requirements for the Degree of Doctor of Physical Therapy from the University Of North Dakota, has been read by the Faculty Advisors under whom the work has been done and is hereby approved.

Graduate School Advisor

Chairperson, Physical Therapy

PERMISSION

Title	Effect of Frontal Plane Foot Position on Lower Extremity Muscle Activation and Limb Positioning in a Single Leg Squat
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ABSTRACT

Purpose: There is a high prevalence of ACL injury in the athletic populations, which can carry out short and long term debilitative effects. Most ACL injuries involve minimal to no contact and female athletes sustain a two to eightfold greater rate of injury than male athletes. Not much research has been conducted to see if foot position directly affects the lower extremity muscles, which could result in altered biomechanics at the knee.

Methods: Twelve subjects 18-30 years old participated in the study. EMG analysis measured differences of muscle contractions for the gluteus maximus, gluteus medius, biceps femoris, rectus femoris, lateral gastrocnemius and anterior tibialis muscles in varied foot positions to include: neutral (control), pronation 5^{o}, pronation 10^{o}, supination 5^{o}, and supination 10^{o}.

Results: When comparing the baseline single leg squat to each of the four test positions (pronation 5^{o}, pronation 10^{o}, supination 5^{o}, and supination 10^{o}) the only significance found was in the anterior tibialis muscle ($p< 0.05$). No significant difference was found in the gluteus maximus, gluteus medius, biceps femoris, rectus femoris, or lateral gastrocnemius muscles in the tested foot positions.

Conclusions: Results of this study show that only the anterior tibialis muscle is affected according to foot position during a single-leg squat. This study suggests that foot position may not have an effect on muscles of the lower extremity and does not play a

major role in non-contact ACL injuries. Many other elements may have affected the results and should be investigated more thoroughly with larger numbers of participants to be more confident of foot position's influence on muscle activity in the lower extremity and role as a possible cause of ACL injury.

CHAPTER I

INTRODUCTION

Lower extremity muscle activity influenced by foot position may alter the position of the knee, thus altering stresses on the anterior cruciate ligament. There is a high prevalence of ACL injury in the athletic populations, which can carry out short and long term debilitative effects. Most ACL injuries involve minimal to no contact and female athletes sustain a two- to eightfold greater rate of injury than male athletes.[1] Most ACL injuries result from low velocity, noncontact, deceleration injuries and contact injuries with a rotational component.[2] There has been a huge focus on noncontact ACL injuries in team sports, but the mechanism of ACL injuries remains unclear.[3,4] Common situations that lead to non-contact injuries include: change of direction or cutting maneuvers combined with deceleration, landing from a jump in or near full extension, and pivoting with knee near full extension on a planted foot.[5] A number of the reasons thought to lead to the ACL's susceptibility to injury include anatomical, biomechanical, hormonal, and neurophysiological, as well as many others. Through varying foot positions during a single leg squat, this study explored a possible component to the mechanism of ACL injuries. This will benefit athletes and those working with the athletic population to prevent ACL injury through interventions that reduce the risk of sustaining such an injury.

CHAPTER II

LITERATURE REVIEW

According to the literature there may be numerous contributors to ACL injuries and the mechanism of these injuries is not well understood. A large extent of the current literature is evaluating the neuromuscular effects that may contribute to ACL injury and/or risk. Although females are at a much higher risk of attaining an ACL injury, many of the researchers can't agree as to why this may be. However, researchers do agree that women are typically found to land with greater peak knee abduction angles than males.[6] Since an increase in the knee abduction angle increases the load on the ACL, it would be reasonable to conclude that this contributing factor could be more of an influence in females in comparison to males. Also, multiple studies have investigated to see if the gender difference in an ACL injury is caused by differences in knee and hip flexion in landings. The differences in muscle strength between genders could result in different landing patterns, thus subjecting the athlete to force differences across the knee joint.[7-11]

In a video analysis study of ACL injuries in basketball players, female players landed with significantly less knee and hip flexion and had a 5.3 times higher relative risk than males of sustaining a valgus collapse.[8] The valgus collapse posture typically included a contralateral pelvic drop, femoral adduction and femoral internal rotation.[10] The study also concluded that women are likely more prone to the anterior quadriceps drawer mechanism upon landing than are men, inducing more stress upon the ACL.[8] This may be attributed to decreased knee flexion, which doesn't allow the hamstrings to be in an effective position to control the anterior drawer mechanism.[12] In return, it may suggest that the smaller knee and hip flexion angles will increase the risk of noncontact ACL injury.[8] Other studies have investigated if the gender difference in ACL injury incidence is caused by differences in knee and hip flexion in landings, with the rationale that women are more extended at the hip and knee during landing, perhaps because of weaker musculature, than are men.[9,10,13,14] Females demonstrate lesser hamstring stiffness compared to males in response to standardized loading conditions, indicating a compromised ability to resist changes in length associated with joint perturbation, and potentially contributing to the higher female ACL injury risk.[10] Boden et al[15] hypothesized that a vigorous quadriceps contraction on an extended knee was the main cause of the excessive ACL force. Although several laboratory studies have supported this theory,[7,14] some studies also found no differences.[13]

In studies pertaining to landing techniques and foot positioning, it was found that a rear-foot landing technique created more ankle dorsiflexion and less knee flexion than did the other techniques such as forefoot landing. A decreased knee flexion angle combined with the knee abductor moment, during the rearfoot landing technique, can create higher stress and strain on the ACL.[16] There was a lack of gender differences in these studies, which may suggest that ACL injuries might not be related solely to gender but may instead be associated with the landing technique used and, as a result, the way each individual absorbs jump-landing energy.[16,17] In a study conducted by Chappell et al,[18] it was found that knee and hip motion patterns as well as quadriceps and hamstring activation patterns exhibited significant gender differences. They concluded that lower extremity motion patterns during landing of the stop-jump task are preprogrammed before landing. Female subjects prepared for landing with decreased hip and knee flexion at landing, increased quadriceps activation, and decreased hamstring activation, which may result in increased ACL loading during the landing of the stop-jump task and the risk for noncontact ACL injury.[18]

Anatomy of the lower extremity is also thought to play a major factor in ACL injury. The ACL controls anterior movement of the tibia and inhibits extreme ranges of tibial rotation. The ACL consists of 2 major bundles, the posterolateral bundle (PL) and the anteromedial bundle (AM). The component ACL bundles are named based on their tibial insertion.[19] Forces transmitted through ACL bundles vary with knee-joint position. In a cadaveric study, the greatest forces transmitted through the AM bundle were at 60 and 90 degrees of flexion. The force was greatest for the PL bundle at full extension.[20]Another study using cadaveric knees found that the PL bundle handled more force overall than the AM bundle in response to anterior tibial loads, whereas the in situ forces in the AM bundle remained relatively constant and unaffected by the changes in flexion angle and anterior tibial load force.[21] In addition, intercondylar notch width was found to be a predictor of ACL injury. Notchwidth index (NWI) is a ratio of intercondylar notch width to femoral condyle width.[22, 23] A study conducted by Lund-Hanssen et al, [24] calculated NWI from measurements taken from x-ray films in a unilateral ACL deficient sample and found NWI to be typically smaller in the injured knee compared to the non-injured knee.[22, 23] However, it has been argued that it is the ACL size rather than notch size that is the important risk factor for ACL injury.[25]

Joint laxity has been a common anatomical feature related to ACL rupture incidences. Clear laxity differences have been observed between males and females, with females often displaying greater genu recurvatum,[26,27] anterior knee laxity,[28-32] and general joint laxity.[33-35] Females are also reported to have 25% to 30% greater frontal-plane and transverse-plane laxity[36-39] and less torsional stiffness[36,40,41] than males. A number of factors contribute to knee joint laxity including hormones, neuromuscular control, and other anatomical structures. Of particular interest here is the role of other structures surrounding the knee. A cadaveric study retrieved the effects of the iliotibial band, capsular ligaments and the medial and lateral collateral ligaments on passive knee joint laxity.[22,42] It showed that the ACL provided most resistance against anterior tibial translation, and those surrounding structures acted as secondary restraints.[22,42] However, the relative contribution of each of those secondary restraints did not differ significantly among each other, and the individual contribution of each structure was minimal.[22] Also, sex hormone (eg, estrogen, testosterone, relaxin) receptors have been found on the human ACL as well as in skeletal muscle, suggesting that hormone signaling may influence ACL susceptibility to injury.[43-46] It has been found that the likelihood of suffering an ACL injury is not evenly distributed across the menstrual cycle in women; instead, the risk of suffering an ACL disruption is greater during the preovulatory phase of the cycle than during the postovulatory phase.[47-52] During the preovulatory phase, hormone levels change dramatically, falling to their lowest levels with the onset of menses and rising rapidly near ovulation. This large hormone swing might contribute to the increase of sustaining an ACL injury during that time period.[47]

Approximately 80% of all ACL tears are noncontact injuries, which may suggest that a high percentage of tears could be avoided through prevention programs.[24,53,54] Randomized controlled studies have shown that proprioceptive training can improve landing mechanics.[22,55,56] Additionally, prospective cohort studies have revealed lower ACL injury rates in cohorts that have undergone proprioception training.[22,57,58] ACL injury prevention programs usually target high-risk groups, such as young female athletes, and aim to improve dangerous motion patterns.[24] There is strong evidence in support of a significant effect of ACL injury prevention programs. Sadoghi et al,[24] found a 52% reduction in the risk of an ACL tear in female athletes but an 85% reduction in male athletes; however, there are no specific types of prevention programs that are more beneficial than others at this time according to literature.[24]

In a study looking at how foot placement modifies kinematics and kinetics during drop jumping, it was found that foot placement while landing during a drop jump clearly modifies the magnitude and distribution of power production. It was found that torque was increased in the ankle, knee, and hip joint, which could also be responsible for the greater power production in forefoot landing when compared with that in heel toe landing.[59] Power production will elicit greater muscle activity at the knee and may contribute to ACL injury if a muscle imbalance is present. Muscle fatigue has also been an attributed factor to increased risk for ACL injury. One study found a significant increase in initial contact hip extension and internal rotation motion, and in peak stance knee abduction and internal rotation and ankle supination angles. They also found that fatigue-induced increases in initial contact hip rotations and in peak knee abduction angles were also significantly more pronounced during unanticipated compared to anticipated landings.[60] With limited research on the foot position effect on an ACL injury, our study may be pertinent to providing more potential causes of the mechanism of injury.

CHAPTER III

METHODS

This study is designed to examine the relationship between frontal plane foot position and lower extremity muscle activation in people while performing a single leg squat. This study will determine if foot position influences muscle activation in the lower extremity, and, therefore, may contribute to the risk of ACL injury. The activity of the lower extremity muscles in various foot positions during a single-leg squat will be measured using EMG analysis of the muscles influencing the knee. To start, the paticipants filled out a data sheet that stated their previous injury/surgery profile, height, weight, and current stage of menstrual cycle (if applicable), all of which addressed possible links to ACL injury risk. After obtaining the participant's history, instructions were given to kick a ball with whichever leg was instinctual. This was done three times and the stance leg that was used while kicking determined the leg used to collect data for the single leg squat.

To collect the data for the single leg squat, disposable electrodes were placed on the skin over each of the subject's lower extremity muscles being tested. The six muscles tested included gluteus maximus, gluteus medius, quadriceps femoris, biceps femoris, anterior tibialis, and lateral gastrocnemius. The skin over which the electrodes were placed was prepared by shaving any hair from the area, lightly abrading the skin with fine sandpaper, and then cleaned with rubbing alcohol. The skin preperation was done to ensure best electrical conductance from muscle to the electromyographic (EMG) equipment. Once the electrodes were in place and the subject was connected to the wireless EMG analysis equipment, instrucutions were given on how to do various LE movements to ensure proper EMG readings. The EMG recording device measures the electrical activity of the muscles during muscle action. Subjects performed a barefoot single-leg squat (45 to 60 degrees knee flexion) on the dominant stance leg 5 times, then returned to standing erect while weight bearing as EMG activity was recorded for each of 5 different foot positions.

After explaining the procedure to the subject, participants were told to choose at random four cards. Each card had a specified foot position associated with it in which the subject would perform in the order that the card was picked. Randomnization was done in this regard to make sure each participant's data wasn't biased due to fatigue or muscle adaptation, which may have occurred with sequencing the postions in a certain order. This allowed for a radomized and variable sequence in which the participants performed each position for the single leg squat. Before recording data for each position, participants were instructed to practice everytime positions were changed. The subjects performed 5 repetitons in each position, with data collection resetting after each position. Muscle activity was reported as a percentage of a single leg stance position on a flat (neutral) board. This was used as our reference measure and then we were able to compare data from the inclines against the EMG activity while on the neutral board. The foot positions tested included a flat surface with the foot in a relaxed position, promoting a neutral foot position; standing on a 5- and 10-degree medial to lateral incline, promoting a pronated foot position (foot arch height lowered); and standing on a 5- and 10-degree lateral to medial incline, promoting a supinated foot position (foot arch height elevated).

Data analysis of the muscles was performed using SPSS. An alpha level of $p \leq 0.05$ was set to determine significance for all statistical tests. A Repeated Measures ANOVA was used to identify differences among groups**.** Mauchly's Test of Sphericity was assessed to see if the assumption of sphericity had been violated, allowing us to choose sphericity assumed or lower-bound according to epsilon. Following that, post hoc analysis was performed with pairwise comparisons.

CHAPTER IV

RESULTS

Data analysis was performed on each of the six muscle groups (gluteus maximus, gluteus medius, biceps femoris, rectus femoris, lateral gastrocnemius, and anterior tibialis) comparing the baseline single leg squat to each of the four test positions (pronation 5°, pronation 10°, supination 5°, and supination 10°). See Figure 1.

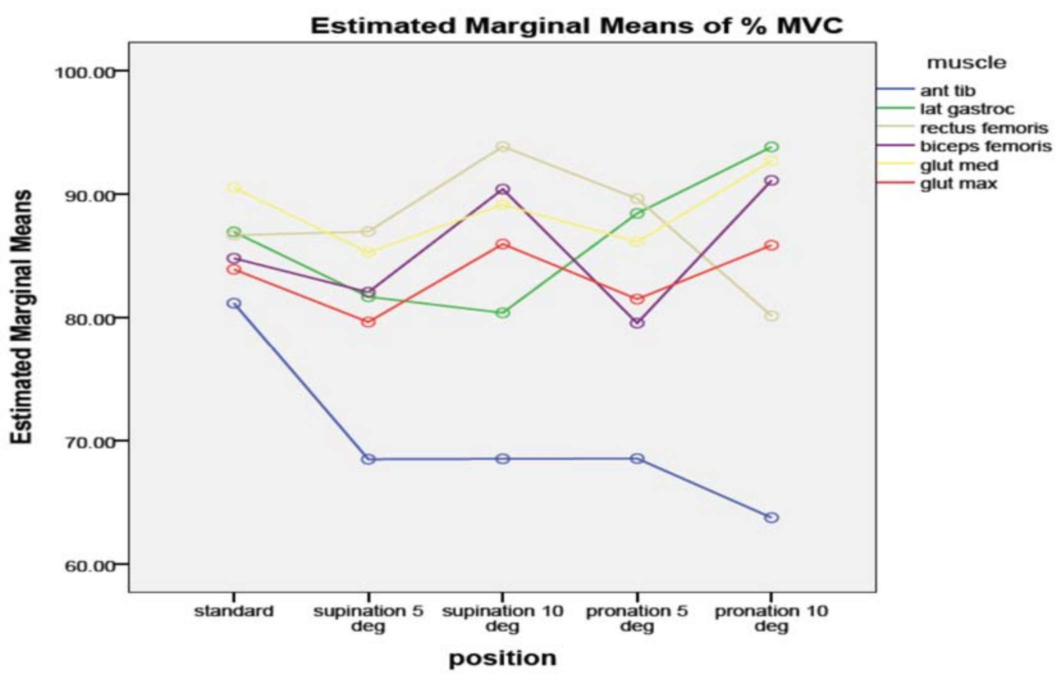

Figure 1.Comparison of muscles means in respective foot positions.

Data analysis shows that there is no significant difference (p > 0.05) between the baseline muscle activity and five of the muscles tested: gluteus maximus, gluteus medius, biceps femoris, rectus femoris, lateral gastrocnemius. See Table 1. The only significant difference (p < 0.05) found was in the anterior tibialis muscle. See Table 1 and 2.

Table 1. Results of statistical inferences of muscles between neutral and tested foot positions.

Muscle	Mauchly's Test of Sphericity	Type III Sum of Squares	df	Mean Square	F	Sig	Partial Eta Squared	Observed Power
Anterior Tibialis	Sphericity Assumed	2033.991	4	508.498	3.600	.013	.247	.835
Lateral Gastroc	Sphericity Assumed	1409.529	4	352.382	1.865	.134	.145	.520
Rectus Femoris	Lower-bound	1198.213	1	1198.213	.936	.354	.078	.143
Biceps Femoris	Sphericity Assumed	1234.722	4	308.681	2.282	.076	.172	.617
Gluteus Medius	Sphericity Assumed	451.759	4	112.940	.655	.627	.056	.197
Gluteus Maximus	Lower-bound	800.261	1	800.261	1.247	.293	.122	.170

The rectus femoris and gluteus maximus muscles used a lower-bounds statistic as the assumption of Mauchly's sphericity was violated (p <0.05).The anterior tibialis, lateral gastrocnemius, biceps femoris and gluteus medius muscles did not violate Mauchly's sphericity (p > 0.05); therefore, the sphericity assumed statistic was used.

Table 2. Statistical analyisis of the anterior tibialis muscle between foot positions.

Muscle Position	Mean	SD	N
Anterior Tibialis Neutral (control)	81.158	12.013	12
Anterior Tibialis Supination (5 degrees)	68.508	13.032	12
Anterior Tibialis Supination (10 degrees)	68.542	23.007	12
Anterior Tibialis Pronation (5 degrees)	68.567	21.123	12
Anterior Tibialis Pronation (10 degrees)	63.783	22.673	12

CHAPTER V

DISCUSSION

The results of the study showed that all of the muscle groups, except for anterior tibialis, had no significant variation in muscle contraction between foot positions during the single-leg squat. Many factors could attribute to the non-significance of the study. One factor that may have influenced our results is that the squatting technique was not standardized among subjects. For example, participants were allowed to squat with their non-weight bearing leg either in front of them or behind them as long as it was not assisting their squat. This could have altered the biomechanics by having the trunk more forward with the non-weight bearing leg posterior or by having the trunk more in line with the weight bearing leg and the non-weight bearing leg more forward. These observed techniques may have allowed for various neuromuscular control tendencies throughout participants. One study showed that single-leg squats performed with a moderate forward trunk lean (~40°) can minimize ACL loads. Also, a moderate vs. minimal forward trunk lean can produce 35% higher hamstring forces throughout the majority of the squat, but only lowers quadriceps forces at knee flexion angles greater than 65°.[61]

Another aspect that wasn't incorporated into our study was the difference between genders. There could be numerous differences between genders that could have influenced the results. Neuromuscular, biomechanical, hormonal, muscular, and joint laxity differences were not taken into effect. Since all of these could influence the results significantly[51,52], future studies should incorporate these factors to see if there is an influence. Studies could also compare each gender respectively to see if there is a possible link to lower extremity muscular difference in foot position related to either gender.

Furthermore, our study did not look into any biomechanical/structural differences between participants. We did not measure any anteversion/ retroversion at the hip, valgus/varus at the knee, or rearfoot and forefoot varus/valgus, all which could feature different biomechanics of the single leg squat and altered activation of certain muscles in the lower extremity. Females have greater mean anterior pelvic tilt, hip anteversion, quadriceps angles, tibiofemoral angles, and genu recurvatum than males, which could influence the muscles in control of those joints and other joint stabilizers.[52] The participant's muscular strength and endurance was not tested prior to the study, which may have led to compensatory muscular activation of the muscles tested as a result of weakness or fatigue of certain similar activated muscles. On the other hand, the muscles that were being tested may have been compensated for by other muscles that weren't included in the study.

Another problematic issue we ran into during the study was in the preparatory stages with each participant. After the electrodes were placed on the participant according to landmarks on the body, conductance of the electrodes was not always optimal. This was especially noted in the gluteus medius and gluteus maximus muscles. This could have been attributable to the area being harder to shave and clean because of clothing covering the areas. Also, electrical noise in that area caused by the clothing could have influenced the EMG readings as well.

This study would benefit by incorporating a greater number of participants. It is difficult to get good insight and statistical significance with low numbers of subjects (12). Our observed power was very low for all of the muscles except anterior tibialis. See Table 1. This may be attributable to the low sample size and high variance that our study displayed. We also had to remove two gluteus maximus data sets before calculating statistics due to the outliers that they produced. This again may be attributed to the possible noise from the electrodes and cords being bunched under clothing or not having the proper conductance of the electrodes.

Although these factors could have played a role in the muscles activation not being influenced by foot position, the results may also show true insignificance. The muscles that were tested may indeed not be influenced by foot positioning. With so many influences to non-contact ACL injuries, foot positioning may not be a major factor contributing to ACL injury incidence.

CHAPTER VI

CONCLUSION

Data for12 subjects aged 18 to 30 years of age was gathered. The data was derived from the means of the gluteus maximus, gluteus medius, biceps femoris, rectus femoris, lateral gastrocnemius, and anterior tibialis muscle contraction differences between a control foot position (neutral) and the 4 other foot postions (pronation 5 degrees, pronation 10 degrees, supination 5 degrees, supination 10 degrees). There was a significant difference found in the anterior tibialis muscle activity between foot positions. This study reports no significant difference among gluteus maximus, gluteus medius, biceps femoris, rectus femoris, and lateral gastrocnemius muscles in the various foot positions.

Research has demonstrated many influences that may attribute to non-contact ACL injury, but our study has found no significance of foot positioning affecting the activation of muscles in the lower extremity. Future research is recommended using a more accurate and less variable measurement for the EMG analysis between muscles as well as a larger sample size. Further investigation into gender and other biomechanical factors within the single-leg squat and foot positioning should be evaluated.

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